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HELICAL GRAFT

This invention relates to grafts.

We have previously proposed that the flow pattern
in arteries including the swirling pattern induced by
their non-planar geometry operates to inhibit the
development of vascular diseases such as thrombosis,
atherosclerosis and intimal hyperplasia.

It is known from WO 95/09585 to provide a vascular
prosthesis comprising a length of generally hollow
tubing having openings at both ends thereof and
including a non-planar curved portion so as to induce
swirl flow in blood flowing through the curved portion.
As explained in that publication, the swirl flow induced
by skewing of the blood flow within the non-planar
curved portion improves flow characteristics and reduces
the potential for deposit build-up and vascular disease
including intimal hyperplasia.

In WO 98/53764, there is disclosed a stent for
supporting part of a blood vessel. The stent includes a
supporting portion around which or within which part of
a blood vessel intended for grafting can be placed so
that the stent internally or externally supports that
part. The supporting portion of the stent is shaped so
that flow between graft and host vessel is caused to
follow a non-planar curve. This generates a swirl flow,
again to provide a favourable blood flow velocity
pattern which reduces the occurrence of vascular
disease, particularly intimal hyperplasia.

In WO 00/32241, there is disclosed another type of
stent, in this case including a supporting portion
around which or within which part of an intact blood
vessel other than a graft can be placed. This
supporting portion can prevent failure of the vessel
through blockage, kinking or collapse. Again, the
supporting portion of the stent is of a shape and/or

orientation whereby flow within the vessel is caused to follow a non-planar curve. Favourable blood flow velocity patterns can be achieved through generation therein of swirl flow within and beyond the stent.

5   Failures in blood vessels through diseases such as thrombosis, atherosclerosis, intimal hyperplasia can be significantly reduced.

Further aspects of how swirl flow is beneficial are explained in the above publications. It is further

10   explained in Caro et al. (1998) J. Physiol. 513P,2P how non-planar geometry of tubing inhibits flow instability.

In certain embodiments of the above publications the artificial or modified natural blood flow tubing is helical or part-helical. In the case of part-helical

15   tubing, the prosthesis or the supported vessel may undergo less than one complete turn of a helix, for example less than one half or less than one quarter of such a turn.

In this specification, the "swept width" of a helix

20   means the outer width of the helix when viewed axially of the helix. In cases where this swept width is relatively wide compared to the width of the tubing itself, the prosthesis or stent may be more bulky than is necessary or acceptable to induce the required swirl

25   flow.

It has been proposed in WO 00/38591 to use internal helical grooving or ridging to induce helical flow. Similar proposals have been made in WO 97/24081 and EP 1127557 A1. However, the use of ribs or grooves in an

30   otherwise cylindrical tube may not reliably induce swirl flow across the entire cross-section of flow. There may be a tendency for the flow nearer to the centre of the tube to follow a linear path, particularly for flows at higher Reynolds numbers. Furthermore, the ratio of the

35   wetted perimeter to the cross-sectional area of a tube is increased by the provision of ridges or grooves. There is a departure from a circular cross-sectional

shape. This may lead to increased flow resistance and a
consequent pressure loss, and damage to blood vessels
and blood cells and the development of pathology.

It is also proposed in WO 00/38591 to use a non-
circular cross-section tube which is twisted. Again,
however, a departure from circularity increases the
ratio of the wetted perimeter to the cross-sectional
area and will have disadvantages.

A further proposal in WO 00/38591 is to provide a
circular-section tube bent into a cork screw shape. It
is usual for the helix of a cork screw to have a clear
gap down the middle, so that this proposed configuration
would have a wide swept width compared to the width of
the tubing, certainly more than two tubing diameters.
The amplitude of the helix would be greater than one
half of the internal diameter of the tubing and there
would be no "line of sight" along the inside of the
tubing. This proposal would therefore be relatively
bulky and unsuitable for certain applications. A
similar proposal is shown in Figure 5 of WO 02/98325,
the tubing having a helix with a large amplitude and
again no "line of sight" along the inside of the tubing.

Various designs of elastomeric arterial graft
prostheses are proposed in GB 2092894. In the version
of Figure 8 of that document, the interior surface is
undulatory or corrugated, with different undulations
either having parallel circumferential paths or joined
in a "spiral" path. The corrugations are proposed as an
alternative to reinforcement for improving the anti-
kinking characteristics of the graft. In the case of
the "spiral" corrugations which appear to be shown in
Figure 8, the angle of the corrugations to the
longitudinal axis is relatively high, of the order of
more than 70°. This is to be expected where the purpose
of the corrugations is to improve anti-kinking or other
structural characteristics, rather than for reasons
relating to the nature of the blood flow through the

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graft. In fact, it is likely that the corrugations
would tend to cause the flow to undergo sharp changes of
direction leading to flow separation and the creation of
stagnant near-wall regions.

5   According to a first aspect of the invention, there
is provided a graft comprising flow tubing having a
tubing portion defining a flow lumen, the flow lumen of
said tubing portion being substantially free of ribs or
grooves, wherein the centre line of the flow lumen

10   follows a substantially helical path with a helix angle
less than or equal to 65°, and wherein the amplitude of
the helix is less than or equal to one half of the
internal diameter of the tubing portion.

       The invention is particularly suitable for *in vivo*

15   tubing, such as vascular prostheses. It is very
suitable for vascular access grafts because (unlike with
tubing having a large amplitude helix relative to its
internal diameter) the tubing can be readily punctured
with a needle to allow blood to be withdrawn and

20   returned for e.g. dialysis.

       The graft according to the invention improves flow
characteristics. As is well known, in the case of
straight tubes, near wall velocities are very low
compared to velocities at the core of the tube, due to

25   the effects of viscosity. In the case of tubes which
are bent in a single plane, the speed of the flow at the
outside of the bend is increased but the speed of the
flow at the inside is retarded further. In both cases,
there is considerable variation in axial velocity across

30   the width of the tube. With the use of a helical tubing
portion according to the invention, a swirl flow is
generated and the axial velocity profile of the flow
across the tubing portion becomes generally more uniform
or "blunter", with the axial velocity of flow at both

35   the outside and inside of the tubing portion being
closer to the mean axial velocity.

       Thus, the flow characteristics are improved by

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causing swirling and a relatively uniform distribution
of axial and near wall velocity. Mixing over the cross
section is also promoted and there is a reduction in the
likelihood of occurrence of flow instability. The

5   avoidance and flushing of stagnant zones is assisted.
There is a reduction in the potential for deposit build
up within and downstream of the graft and the
development of pathology.

In this specification, the amplitude of the helix

10   refers to the extent of displacement from a mean
position to a lateral extreme. So, in the case of the
tubing having a helical centre line, the amplitude is
one half of the full lateral width of the helical centre
line.

15   In the tubing of the first aspect of the invention,
in which the amplitude of the helix is less than or
equal to one half of the internal diameter of the
tubing, there is a "line of sight" along the lumen of
the tubing, unlike in the case of a corkscrew

20   configuration where in effect the helix is wound around
a core (either solid, or "virtual" with a core of air).
We have found that the flow at the line of sight
generally has a swirl component, even though it could
potentially follow a straight path.

25   . For the purposes of this specification, the term
"relative amplitude" of a helical tubing is regarded as
the amplitude divided by the internal diameter. So, in
the tubing of the first aspect of the invention in which
the amplitude of the helical tubing is less than or

30   equal to one half of the internal diameter of the
tubing, this means that the relative amplitude is less
than or equal to 0.5. Relative amplitudes less than or
equal to 0.45, 0.4, 0.35, 0.3, 0.25, 0.2, 0.15 or 0.1
may be preferred in some circumstances. It is however

35   preferred for the relative amplitude to be at least
0.05, more preferably 0.1. This can help to ensure that
the desired swirl flow is induced.

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The relative amplitude may vary according to the use of the tubing and the spatial constraints on its design. It will however be appreciated that by keeping the amplitude less than half the tubing internal diameter a swirling flow may be induced without creating an excessively large device. The "envelope" occupied by the graft can fit into the space available in the surrounding tissue, and even if this envelope is caused to follow a particular path by the local environment in which the graft is located, the desired helical geometry of the flow lumen can be maintained.

The angle of the helix is also a relevant factor in balancing the space constraints on the flow tubing with the desirability of maximising the cross-sectional area available for flow. The helix angle is less than or equal to 65°, preferably less than or equal to 55°, 45°, 35°, 25°, 20°, 15°, 10° or 5°. As with relative amplitudes, the helix angle may be optimized according to the conditions: viscosity, density and velocity of fluid.

Generally speaking, for higher Reynolds numbers the helix angle may be smaller whilst satisfactory swirl flow is achieved, whilst with lower Reynolds numbers a higher helix angle will be required to produce satisfactory swirl. The use of higher helix angles will generally be undesirable, as there may be near wall pockets of stagnant fluid. Therefore, for a given Reynolds number (or range of Reynolds numbers), the helix angle will preferably be chosen to be as low as possible to produce satisfactory swirl. Lower helix angles result in smaller increases in length as compared to that of the equivalent cylindrical tubing. In certain embodiments, the helix angle is less than 20° or less than 15°.

It will be appreciated that in pulsatile flow, the Reynolds number will vary over a range. Typical mean resting arterial blood flow Reynolds numbers are about

100, reaching peak values of two or three times that in pulsatile flow and three to four times the mean during exertion. Therefore the extent to which swirl flow is promoted will vary likewise. Even if there are stagnant
5   flow regions at lower Reynolds numbers, because for example a low helix angle and/or a low relative amplitude has been selected, these will tend to be flushed out during periods of flow when the Reynolds numbers are higher.

10   The tubing portion may be made with substantially the same relative amplitude and helix angle along its length. There may be small variations when the tubing is in use, caused by elongation or contraction of the tubing portion due to tensile loading or caused by
15   torsional loading. However, there may be circumstances in which the tubing portion has a variable helix angle and/or relative amplitude, either to suit the space constraints or to optimise the flow conditions.

For reasons of manufacturing simplicity, it may be
20   preferred for the tubing portion to have a substantially constant cross-sectional area along its length. Again, there may be variations in use caused by loading on the tubing portion.

The helical tubing portion may form just part of
25   the overall length of tubing or it may extend over substantially its entire length. For example, a prosthesis may have a tubing portion with the geometry of the invention over part of its length or over substantially its entire length.

30   The helical tubing portion may undergo a fraction of one complete turn, for example one quarter, one half or three quarters of a turn. Preferably, the helical tubing portion undergoes at least one turn, more preferably at least a plurality of turns. Repeated
35   turns of the helix along the tubing portion will tend to ensure that the swirl flow is generated and maintained.

The tubing, including the tubing portion, may

extend generally linearly. In other words, the axis
about which the centre line of the tubing portion
follows a substantially helical path, may be straight.
Alternatively the axis may itself be curved, whereby the
envelope occupied by the tubing is curved, for example
to produce an "arch" shaped tubing. The bend of the
arch may be planar or non-planar, but should preferably
be such that swirl is maintained and not cancelled by
the geometry of the bend. Thus, for example, a
prosthesis may be generally "arch" shaped (planar or
non-planar), having the geometry in accordance with the
first aspect of the invention, i.e. being in the form of
a tubing portion following a substantially helical path
with a helix angle less than or equal to 65°, and with
an amplitude less than or equal to one half of the
internal diameter of the tubing portion.

In general, the helical centre line of the tubing
portion is defined by the tubing portion itself and is
not due to a branch in the tubing. The tubing portion
may interface with another portion or a branch which may
be planar or non-planar. The interface between the
helix of the tubing portion and the branch will
preferably be such that swirl is maintained, and that
there is not a tendency for the swirl to be cancelled by
the geometry of the branch.

The tubing may if desired comprise a pharmaceutical
coating. Such a coating could be provided to provide
sustained release of the pharmaceutical over a period of
time. So, the blood flow tubing could provide a
pharmaceutical for initial treatment of a disease, and
in the longer term the tubing portion gives a
therapeutic benefit due to the characteristics which it
imparts to the flow.

In the above prior art proposals using multiple
grooves or ridges arranged about the tubing
circumference, or non-circular sections which are
twisted, where the tubing is substantially straight,

then the centre line of the tubing is also straight.
This is unlike the centre line of the tubing portion of
the present invention, in its first aspect, which
follows a substantially helical path. Thus, the tubing
portion of the invention may have a circular cross-
section and thus the smallest possible wetted perimeter
to cross-sectional area ratio, whilst still having the
necessary characteristics to induce swirl flow. Of
course, there may be circumstances in which the tubing
portion of the present invention has a non-circular
cross-section, for example to assist interfacing or
where pressure loss considerations are not significant.

In the proposals of WO 97/24081 and EP 1127557 A1,
the tubing has a single internal rib arranged helically.
This results in the tubing having a centre line which
follows a helical path, but because the rib is provided
in an otherwise cylindrical tube, the amplitude of the
helix is very small, generally having a relative
amplitude appreciably less than 0.05. The generation of
swirl flow, if there is any, is correspondingly limited
and unsatisfactory.

Further concerning the prior art proposals using
grooves or ridges or ribs, it should be noted that
arterial geometry is under normal physiological
conditions non-planar (i.e. curved in more than one
plane in the nature of a helix) and not grooved or
rifled. We have found experimentally that at higher
relevant Reynolds numbers, the flow in a helical (non-
planar) geometry differs from that in a rifled/grooved
geometry, e.g. there is swirling of both near-wall flow
and core flow in the former case. The development of
swirl flow is more rapid than in the case of
rifled/grooved tubing, where swirl flow can take many
tubing diameters to develop. Thus, there is the
expectation that the introduction of the physiological
non-planar geometry (unlike grooved or rifled geometry)
will be beneficial in respect of inhibiting the

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development of pathology.

Because the tubing portion of the invention has a
helical centre line, there is spatial reorganisation of
vortical structures, which results in motion of the core
5   or cores of the axial flow across the section of the
tubing portion, promoting mixing across the cross
section. The swirl inhibits the development of
stagnation and flow separation regions and stabilises
flows.

10   As mentioned, in the case of the prior art
proposals using multiple grooves or ridges or ribs, or
twisted tubes of a non-circular cross-section, the
centre line is straight, not helical. Whilst this can
be expected to stabilise flow at sharp bends, it does
15   not in straight tubes cause spatial reorganisation of
vortical structures, resulting in motion of the core or
cores of the axial flow across the section of the tube.
Thus it does not promote mixing across the cross section
to the same extent as tubing according to the invention.
20   Such mixing may be important in maintaining the mas s
transport and physiological integrity of the blood
vessels.

According to another aspect of the invention, there
is provided a graft comprising flow tubing having a
25   tubing portion defining a flow lumen, wherein the c entre
line of the flow lumen follows a substantially heli cal
path with a helix angle less than or equal to 65°,
wherein the amplitude of the helical centre line is less
than or equal to one half of the internal diameter of
30   the tubing portion, and wherein the amplitude of th e
helical centre line is more than or equal to 0.05 o f the
internal diameter of the tubing portion. The various
other possible features of the graft discussed here in
may be provided in the graft of this aspect of the
35   invention.

The tubing geometry disclosed herein may be used in
various biomedical applications e.g. in various arteries

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(such as in the coronary and renal arteries), in veins,
and in non-cardiovascular applications such as in the
gastro-intestinal (e.g. bile or pancreatic ducts),
genito-urinary (e.g. ureter or urethra) or the
respiratory system (lung airways). Thus, the invention
extends to flow tubing for body fluids other than blood.
In general, the use of the tubing geometry of the
invention can avoid the presence of stagnant regions,
and hence be beneficial.

If tubing is made from flexible material, such as
synthetic fabric, but rather than being formed as a
cylinder is instead formed so that its centre line
follows a substantially helical path, it is in some
circumstances capable of "straightening out", involving
a reduction in the amplitude of the helix and a
corresponding increase in the pitch of the helix and in
the length of the tubing (i.e. axial extension). The
benefits of swirl flow discussed above may then be
reduced or lost.

According to another aspect of the invention there
is provided a graft comprising flow tubing having a
tubing portion, the tubing portion comprising a wall
defining a longitudinally extending flow lumen which is
substantially free of ribs or grooves, the flow lumen
having a centre line following a substantially helical
path, and the wall having a helical portion extending
longitudinally and circumferentially so as to resist
reduction of the amplitude of the helical centre line.

A helical portion according to this aspect of the
invention can therefore help to maintain the desired
amplitude of the helical centre line, and hence maintain
the desired swirl fluid flow characteristics.

There are a number of situations where tubing could
be subjected to "straightening out" effects tending to
cause helical amplitude reduction. These include
internal pressurisation by a fluid, for example in
response to arterial pressure, or axial extension if the

graft is used in the vicinity of a joint, or a
combination of the two. Although there may still be a
reduction in amplitude when the tubing is subjected to
such straightening out forces, the amount of this

5   reduction is less than would be the case without the
helical portion.

In general, the helical portion will have a lower
extensibility as compared to adjacent portions of the
tubing. It will normally have the same pitch as the

10   helical centre line of the flow lumen so as to conform
therewith.

The invention is particularly suitable for *in vivo*
tubing, such as vascular prostheses, and for vascular
access grafts.

15   The longitudinal cavity of the tubing wall itself
provides the lumen for body fluid flow. The fluid may
then act directly on the tubing wall and create the
internal pressurisation tending to straighten out (i.e.
to increase the pitch and reduce the helical amplitude)

20   the tubing and hence the lumen. This is resisted by the
helical portion.

The helical portion may be thicker in the radial
direction than adjacent portions of the tubing wall.
This is a way of achieving the result of the helical

25   portion having lower extensibility than the adjacent
portions. Alternatively or additionally, the helical
portion may be made from a material different from that
of adjacent portions of the tubing wall.

In order to avoid excessive lateral bulk, the

30   amplitude of the helical centre line of the tubing
longitudinal flow lumen, once internally pressurised,
may be less than or equal to one half of the internal
diameter of the tubing. It is expected that any
straightening out of the tubing, and hence reduction in

35   the relative amplitude, when the tubing is in use will
not be significant, because of the presence of the
helical portion.

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The various other possible features of the graft
discussed herein (such as in relation to the amplitude
of the helical centre line, the helix angle, the
constance or variation of the amplitude or the helix
5   angle, the number of turns and so forth) may be provided
in the graft of this aspect of the invention.

The invention also extends to methods of
manufacturing grafts.

According to another aspect of the invention,
10   therefore, there is provided a method of making a graft,
the method comprising positioning a generally tubular,
flexible wall adjacent to a further flexible member,
twisting the tubular flexible wall and the flexible
member around each other, and causing the tubular
15   flexible wall to retain, at least partly, the twisted
shape.

By using a flexible member, the amplitude of the
twisted tubular wall can be kept desirably small, so as
to form tubing without excessive lateral bulk. If the
20   tubular wall were instead twisted round a rigid member,
then it would adopt a corkscrew configuration, in effect
a helix round a core provided by the rigid member. If
the tubular wall retained that shape when the rigid
member is removed, it would then have a core of air and
25   be laterally bulky.

In general, the tubular wall formed by twisting
round a flexible member will define a longitudinally
extending cavity having a centre line following a
substantially helical path. The relative amplitude of
30   helical tubing formed by the method discussed is
preferably less than or equal to 0.5. The tubing may
have relative amplitudes, helix angles, cross-sectional
shapes, number of turns etcetera as discussed above in
relation to the other aspects of the invention.

35   It will generally be undesirable for the cross-
sectional shape of the tubular wall to be distorted, for
example flattened, during twisting. Therefore, the

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tubular wall may be reinforced to assist it in
maintaining its cross-sectional shape during twisting
with the flexible member. The reinforcement may be
integral with or adherent to the tubular wall, for
5   example comprising a helical winding with a large helix
angle, as is known for example from GB 2298577.
Alternatively, or additionally, it may be desirable to
provide reinforcement in the form of internal support
for the tubular wall during twisting of the tubular
10   wall. A flexible rod or tube or a spring may be
inserted into the tubular wall to provide internal
support and removed after the desired geometry has been
at least partly retained.

A preferred cross-sectional shape of the
15   longitudinally extending helical cavity is substantially
circular. If reinforcement is provided, it may then
help the tubular wall to keep to this shape.

The further flexible member may for example be
another generally tubular, flexible wall. This may be
20   reinforced if necessary to assist it in maintaining its
cross-sectional shape.

The step of at least partially retaining the
twisted shape may comprise thermosetting the tubular
flexible wall and allowing it to cool.

25   It has been found that tubing made by the above
method need not have a helical portion extending
longitudinally and circumferentially of the wall to help
resist reduction of the amplitude of the helical centre
line. For example, tubing made of ePTFE (expanded
30   polytetrafluoroethylene) and of a conventional type for
use as vascular prostheses has been found generally to
retain the desired geometry without the need for a
helical portion acting as "reinforcement". However, for
tubing made of other biocompatible materials, in view of
35   the potential straightening out effects on tubing having
a twisted shape when the tubing is in use, it may be
preferred to provide the tubing flexible wall with a

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helical portion extending longitudinally and
circumferentially and for assisting in retaining the
twisted shape. In order that the helical portion will
complement the twisted shape achieved by the twisting
5   step, it is preferably positioned to lie adjacent to the
flexible member (for example in contact therewith).

According to another aspect of the invention, there
is provided a method of making a graft, the method
comprising providing a helical mandrel having a centre
10   line following a substantially helical path, providing a
generally tubular, flexible wall having a longitudinally
extending cavity, positioning the tubular wall adjacent
to the helical mandrel to cause the longitudinally
extending cavity to have a centre line following a
15   substantially helical path, and causing the tubular wall
to retain, at least partly, the shape with the
longitudinally extending helical cavity.

With this manufacturing method it is not necessary
to use a flexible member as the mandrel and the helical
20   mandrel may be substantially rigid. This enables the
geometry of the helical mandrel to be fixed in advance
of its use with the tubular wall to make the graft, so
facilitating consistent production of grafts to a
predetermined specification.

25   Preferably the helical mandrel extends
longitudinally and circumferentially around a
cylindrical space which defines a core of the helical
mandrel, and the outside diameter of the tubular wall is
greater than the diameter of the core of the helical
30   mandrel.

The tubular wall may be reinforced to assist it in
maintaining its cross-sectional shape. The
reinforcement may be integral with or adherent to the
wall. Alternatively, or additionally, the tubular wall
35   may be reinforced by a removable internal support.

The method is suited to a continuous production
process. The tubular wall may be fed to one end of the

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helical mandrel and, following deformation to the desired shape, it may separate from the helical mandrel at the other end thereof. Preferably, therefore, the tubular wall and the helical mandrel are moved in the longitudinal direction relative to each other.

The graft made by the above method may comprise the various other possible features of grafts discussed herein, such as in relation to the amplitude of the helical centre line, the helix angle, the constance or variation of the amplitude or the helix angle, the number of turns and so forth.

According to a further aspect of the invention, there is provided a method of making a graft, the method comprising providing a mandrel, providing a generally tubular, flexible wall having a longitudinally extending cavity, winding the tubular wall around the mandrel to extend circumferentially and longitudinally thereof so as to cause the tubular wall to define a first shape in which its longitudinally extending cavity has a centre line following a substantially helical path, setting the tubular wall, and separating the tubular wall from the mandrel so as to allow the amplitude of the helical centre line to reduce whereby the tubular wall adopts a second shape in which the amplitude of the helical centre line is less than or equal to one half of the internal diameter of the tubular wall.

This aspect of the invention allows a straight and generally rigid mandrel to be used, without creating a graft of excessive lateral bulk. Preferably, the mandrel comprises guide means to aid the winding of the tubular wall around the mandrel. Such a guide means can be used to ensure that grafts are made to the same helix angle each time the mandrel is used.

The setting of the tubular wall is preferably a thermosetting step. If the material of the tubular wall is ePTFE, for example, this will adopt a first shape and then, upon separation from the mandrel, adopt the second

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shape with a reduced helical amplitude.

As with other methods described herein, it may be desirable to reinforce the tubular wall to assist it in maintaining its cross-sectional shape.

5   According to a further aspect of the invention, there is provided a method of making a graft, the method comprising arranging an elongate member helically along a generally tubular, flexible wall so that the elongate member extends longitudinally and circumferentially of

10   the tubular wall, tensioning the elongate member to cause the wall to define a longitudinally extending cavity having a centre line following a substantially helical path, and causing the wall to retain, at least partly, the shape with the longitudinally extending

15   helical cavity.

The helically arranged elongate member thus serves to deform the tubular wall to the shape with the longitudinally extending helical cavity. It may also form the helical portion of the tubing for resisting

20   reduction of the amplitude of the helical centre line, i.e. to help it retain its shape. The elongate member may advantageously therefore serve a dual function and simplify manufacture.

As with the previously described manufacturing

25   methods, it will generally be undesirable for the cross-sectional shape of the tubular wall to be distorted, for example flattened, during tensioning. Preferably, therefore, the tubular wall is reinforced to assist it in maintaining its cross-sectional shape during

30   tensioning of the elongate member. The reinforcement may be integral with or adherent to the tubular wall, for example comprising a helical winding with a large helix angle, as is known for example from GB 2298577. Alternatively, or additionally, it may be desirable to

35   provide reinforcement in the form of internal support for the tubular wall during tensioning of the elongate member. A flexible rod or tube or a spring may be

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inserted into the tubular wall to provide internal
support and removed after the desired geometry has been
at least partly retained.

A preferred cross-sectional shape of the
longitudinally extending helical cavity is substantially
circular. If reinforcement is provided, it may then
help the tubular wall to keep to this shape.

The step of at least partially retaining the
tubular wall in a shape with a longitudinally extending
helical cavity is preferably a thermosetting step.
Preferably therefore the materials of the tubular wall
and the elongate member are such as to permit
thermosetting of the tube in the desired shape. It is
preferred for the elongate member to be such that it
retains its tension when heated, i.e. it does not soften
or melt to the extent that it allows the tubular wall to
straighten out. The elongate member preferably also
bonds to the tubular wall when heated, for example by
melting. Then, when cooling takes place the elongate
member is bonded to the tubular wall and holds it in the
shape with the longitudinally extending helical cavity.
An elongate member made of a biocompatible polymer, e.g.
polypropylene, heated to just above its melting point
for an appropriate time can provide both the tension
retention and the bonding properties.

Alternatively, the elongate member may be of
composite construction, including a first material which
retains tension when heated and a second material which
bonds to the tubular wall. The elongate member may then
comprise a tensile element, such as a metal wire, in a
sleeve for bonding to the tubular wall. The sleeve may
be made of a biocompatible polymer which can soften
sufficiently when heated to bond to the tubular wall.
The tensile element may if desired be removed from the
sleeve after the tubular wall has set in the desired
shape. This may be of benefit if the tensile element is
not biocompatible.

According to another aspect of the invention, there
is provided a method of making a graft, the method
comprising providing a generally tubular wall with a
helical portion extending longitudinally and

5   circumferentially, the helical portion being less
extensible than adjacent portions of the wall, and
radially expanding the wall, whereby the helical portion
causes the wall to define a longitudinally extending
cavity having a centre line following a substantially

10   helical path.

It is preferred in the above method to cause the
tubular wall to retain, at least partly, the shape with
the longitudinally extending helical cavity. This may
be achieved for example by thermosetting.

15   Certain preferred embodiments of the invention will
now be described by way of example and with reference to
the accompanying drawings, in which:

Figure 1 is an elevation view of a tubing portion
in accordance with the invention;

20   Figure 2 is a perspective view of a vascular graft;

Figure 3 is a perspective view of another vascular
graft;

Figure 4 is a perspective view of a vascular graft;

Figure 5 is a view of an experimental balloon;

25   Figure 6 is a view of a vascular graft twisted with
a flexible member during manufacture;

Figure 7 is a view of part of the vascular graft of
Figure 6, to an enlarged scale;

Figure 8 is a view of the vascular graft made by

30   the method shown in Figures 6 and 7;

Figure 9 is a view of another vascular graft made
by the same method;

Figures 10a and 10b are views illustrating another
method of manufacturing a graft;

35   Figures 11a and 11b are views illustrating another
method of manufacturing a graft;

Figures 12a to 12e are views illustrating a method

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of manufacturing a graft;

Figures 13a and 13b are views illustrating another method of manufacturing a graft;

Figure 14 shows elevation views of tubing portions
5   used in experiments; and

Figure 15 shows elevation views of tubing portions used in further experiments.

The tubing portion 1 shown in Figure 1 has a circular cross-section, an external diameter D_E, an
10   internal diameter D_I and a wall thickness T. The tubing is coiled into a helix of constant amplitude A (as measured from mean to extreme), constant pitch P, constant helix angle θ and a swept width W. The tubing portion 1 is contained in an imaginary envelope 20 which
15   extends longitudinally and has a width equal to the swept width W of the helix. The envelope 20 may be regarded as having a central longitudinal axis 30, which may also be referred to as an axis of helical rotation. The illustrated tubing portion 1 has a straight axis 30,
20   but it will be appreciated that in alternative designs the central axis may be curved. The tubing portion has a centre line 40 which follows a helical path about the central longitudinal axis 30.

It will be seen that the amplitude A is less than
25   the tubing internal diameter D_I. By keeping the amplitude below this size, the space occupied by the tubing portion can be kept relatively small, whilst at the same time the helical configuration of the tubing portion promotes swirl flow of fluid along the tubing
30   portion.

Figure 2 shows a prosthesis 10 comprising a length of hollow tubing having an inlet 2 at one end and an outlet 3 at the other end. A generally helical tubing portion 1 is provided at the outlet 3 thereof. The
35   prosthesis has inlet 2a and outlet 3a flaps at its ends which have been surgically fastened by suturing to regions of an artery remote from a blockage 7 in the

artery, the prosthesis thus acting as an arterial bypass
graft. It could also be surgically connected between an
artery and a vein so as a vascular access graft for e.g.
renal dialysis.

5   Blood from the circulatory system can flow from the
inlet 2 to the outlet 3 along a hollow interior or lumen
4. The helically formed tubing portion 1 is disposed
adjacent to the outlet 3. Its non-planar curvature
induces a swirl to the flow to improve circulation by
10   rendering the distribution of wall shear stress
relatively uniform and suppressing flow
separation and flow instability, and as a
result inhibiting the development of vessel pathology.
The swirl flow may also resist the build up of intimal
15   hyperplasia at the join and downstream of the join with
the vein or artery. The tubing can be made of suitable
bio-compatible material and such materials are
commercially available and known to those skilled in the
art. In order to maintain the tubing open and prevent
20   collapse or kinking it is possible to use a stent or
other structural support of plastic, metal or other
material internally, externally or integral to the wall
of the tubing.

          It will be seen that the prosthesis 10 in Figure 2
25   is generally arch shaped. This arch may itself be
provided in a single plane. If the arch is non-planar
then this will also tend to induce swirl flow and it
will be desirable to ensure that the swirl flow induced
by the non-planar arch is in the same direction as that
30   induced by the helical tubing portion 1.

          The arrangement of Figure 3 is similar to that of
Figure 2, except that the helically formed tubing
portion 1 extends substantially the full length of the
prosthesis 10. This type of arrangement may simplify
35   manufacture as the tubing could be made in a continuous
length which simply has to be cut to appropriate shorter
lengths to form prostheses.

Part of the envelope 20 within which the tubing
portion 1 is defined is shown in Figure 3. The swept
width W defines the width of the envelope. The
longitudinal axis 30 of the envelope is curved, the
5   tubular portion being arch shaped. The centre line 40
follows a helical path about the axis 30.

The vascular graft 10 shown in Figure 4 has a
substantially circular cross-section. The tubing is,
coiled into a helix of constant amplitude A (as measured
10   from mean to extreme), constant pitch P, constant helix
angle θ and a swept width W. The tubing 1 is contained
in an imaginary envelope 20 which extends longitudinally
and has a width equal to the swept width W of the helix.
The envelope 20 may be regarded as having a central
15   longitudinal axis 30, which may also be referred to as
an axis of helical rotation. The illustrated tubing 1
has a curved axis 30. The tubing has a centre line 40
which follows a helical path about the central
longitudinal axis 30.

20   The tubing 1 has a helical portion 6 extending
longitudinally and circumferentially with the same pitch
as pitch P of the helical centre line 40. The helical
portion 6 consists of a strip of material secured to the
wall 62 of the tubing 1.

25   The tubing 1 has an inlet 2 at one end and an
outlet 3 at the other end. The tubing has inlet 2a and
outlet 3a flaps at its ends which have been surgically
fastened by suturing to regions of an artery 8 remote
from a blockage 7 in the artery, the graft 10 thus
30   acting as an arterial bypass graft. It could also be
surgically connected between an artery and a vein so as
to serve as a vascular access graft for e.g. renal
dialysis.

Blood from the circulatory system can flow from the
35   inlet 2 to the outlet 3 along a hollow interior or lumen
4 of the graft 10. It operates in a manner similar to
the graft of Figure 3, having a non-planar curvature

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and resist the development of pathology. The swirl flow
may also resist the build up of intimal hyperplasia at
the join and downstream of the join with the vein or
artery.

5   The tubing 1 may be made of various materials.
Suitable bio-compatible materials are commercially
available and known to those skilled in the art. One
suitable material is polyester. A knitted polyester
yarn such as polyethylene terephthalate, known as Dacron
10   (trade mark) is a particular example. The helical
portion may be made of the same material or another
material, such as polypropylene. The helical portion,
rather than being a separate strip secured to the wall
62 of the tubing 1, may be an integral part thereof, for
15   example by being knitted or stitched in to the wall.

The tubing 1 may be made of various materials.

Figure 5 shows the result of an experiment carried
out on a toy balloon 55. The balloon was of the
elongated type. It was supported, without being
inflated, on a cylindrical rod and a plastic strip 51
20   cut from another balloon was glued onto the outside of
the supported balloon to form a longitudinally and
circumferentially extending helical strip 6. A straight
line 50 was drawn along the balloon. After the glue had
set, the balloon was inflated and the inflated balloon
25   is shown in Figure 5.

It will be seen that the inflated balloon 55 has a
helical lumen. As with the tubing for fluid flow, it
has a helical centre line 40, which follows a helical
path about a longitudinal axis 30. The longitudinal
30   axis is at the centre of an imaginary cylindrical
envelope 20 within which the balloon is contained. The
amplitude A of the helix is shown in Figure 5.

It will be noted that after inflation the straight
line 50 adopts a wave shape which remains consistently
35   along the same side of the balloon, so that the entire
line 50 remains visible in the elevation view of Figure
5.

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The balloon of Figure 5 starts as a cylindrical membrane with a helical portion which is of greater (in this case double) wall thickness than the rest of the balloon. During inflation the thicker helical portion
5 will tend to resist extension in all directions, including circumferential and longitudinal directions, thereby influencing the shape of the expanded balloon. Instead of adopting the normal cylindrical shape, the balloon forms a shape with a helical centre line 40.

10 The balloon is internally pressurised in a manner to some extent analogous with the internal pressurisation of the tubing of the preferred embodiments of the invention. The helical portion causes what would otherwise be a cylindrical shape to
15 adopt and maintain helical geometry. A similar effect is obtained by the helical portion of the tubing for body fluid flow, wherein the helical portion tends to help the tubing maintain its helical longitudinal cavity, i.e. to resist "straightening out".

20 A tubing having a wall defining a longitudinally extending cavity having a centre line following a substantially helical path was manufactured as follows.

A pair of flexible cylindrical tubes made from polyester were internally supported by insertion of
25 respective closely fitting coiled springs. The two supported tubes were then positioned adjacent to each other and twisted around each other. The pair of tubes were thermoset in the twisted configuration by immersion in hot water followed by removal and cooling. The tubes
30 were separated and the coil springs removed. The internal geometry of each tube so formed consisted of a longitudinally extending cavity having a centre line following a substantially helical path. One of the tubes was subjected to internal pressurisation by
35 insertion of a cylindrical balloon which was then gently inflated. Because of the flexible nature of the material forming the tube, the effect of the internal

pressurisation was to straighten out the helix, in that the pitch was increased and the amplitude decreased.

Such a straightening out effect is however resisted by the use of a helical portion applied to the tube, as
5   described herein. The helical portion is applied to each of the tubes before they are deformed and thermoset as described above. During the step of twisting the two tubes around each other, they are positioned so that their respective helical portions lie in contact with
10 each other.

A similar method was used to manufacture another tubing having a wall defining a longitudinally extending cavity with a centre line following a substantially helical path. In this case, the tubing was made of
15 expanded polytetrafluoroethylene (ePTFE). Biocompatible tubing of this type is available for use as vascular prostheses, for example from Vascutek Limited or Boston Scientific Corporation.

Referring to Figures 6 and 7, a length of ePTFE
20 tubing 1 was internally supported by insertion of a length of silicone rubber tubing 70. A length of polyvinyl chloride (PVC) tubing 71 was internally supported by insertion of a closely fitting coiled spring. The two supported tubes were positioned
25 adjacent to each other and twisted around each other. The support tube 70 was clamped at each end by respective clamps 73, these clamps also serving to clamp the ends of the PVC tube 71. The internally supported, twisted and clamped tubes were placed in an oven at
30 180°C for 5 minutes and then cooled by immersion in water at room temperature. The tubes were separated and the support tube 70 was removed from the tubing 1. The tubing was thermoset in a twisted configuration, as seen in Figure 8. Although the amplitude of the helix was
35 reduced compared to the amplitude during the heating step, the tubing had the desired longitudinally extended cavity with a centre line following a substantially

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helical path.

A test was carried out on the tubing 1 to investigate its ability to maintain its helical geometry. One end was clamped and the other end was connected to a water supply at a pressure head of 1.5 metres (roughly equal to blood pressure). It was observed that the helical geometry was maintained after 24 hours.

Figure 9 shows another length of ePTFE tubing manufactured using the above method. In this case the tubing 1 used at the start was of the armoured type, having an external helical winding 74 with a large helix angle (close to 90°). This type of tubing is used in prostheses subject to external bending forces, for example going across joints such as the knee, and the helical winding serves to help maintain a circular cross-section. It will be noted that such armoured tubing was also successfully modified to have a longitudinally extending cavity with a centre line following a substantially helical path.

In an alternative manufacturing method, only one tube, rather than two, is used. The method is described with reference to Figures 10a and 10b. An elongate member, in the form of a thread 101, is helically wound round an initially cylindrical tube 1. As seen in Figure 10a, the thread 101 is arranged helically along the tubing so as to extend longitudinally and circumferentially thereof. The thread is tensioned and causes the tube to distort helically, such that its longitudinally extending cavity has a centre line following a substantially helical path. The pitch is dictated by the pitch of the winding of the thread. The amplitude is dictated by the tension on the thread. The tension, and hence the helical deformation, is maintained by securing the ends of the thread, for example to a suitable rig. The deformed tube is then heated so as to thermoset and so as to soften the thread

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sufficiently for it to bond to the tube. The thread
therefore serves the purposes first of creating the
helical geometry during the tensioning step, and later
of helping to retain that geometry when the tube is used
5   and internally pressurised by e.g. arterial pressure.
As with other methods described herein, the tubing may
be externally or internally supported during this
process.

In a preferred method a knitted polyester yarn such
10   as polyethylene terephthalate, known as Dacron (trade
mark), is a suitable material for the tube, whilst the
elongate member may be polypropylene. The tube may be
externally supported with helically wound (with a very
large helix angle, close to 90°) polypropylene. With
15   these materials the heating step is carried out by
heating the tube and tensioned thread in an oven at
140°C.

In another alternative manufacturing method using
only one tube, the tube is initially cylindrical, with a
20   helical portion extending along its wall. The method is
described with reference to Figures 11a and 11b. In
this method, tubing 1 is provided with a reinforcing
strip 51 adhered to its outside surface so as to extend
longitudinally and circumferentially of the tubing. An
25   inflatable device 55 is located inside the tubing. The
inflatable device is inflated in order to expand the
tubing. During this process the helically arranged
strip 51 causes the tubing to expand to a shape having a
longitudinal, helical cavity, as seen in Figure 11b.
30   The tubing adopts the helical geometry in the same
manner as the balloon shown in Figure 5. The tubing is
thermoset in this condition and allowed to cool, in
order to retain the desired helical shape. The material
of the inflatable device 55 is chosen to withstand the
35   elevated temperature required to thermoset the tubing.

The helical portion, in the form of strip 51, thus
serves the purposes first of creating the helical

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geometry during the inflation step, and later of helping
to retain that geometry when the tube is used and
internally pressurised by e.g. arterial pressure.

5   Another method of making a graft is described with
reference to Figures 12a to 12e. This method involves
the use of a helical mandrel.

Figure 12a is a schematic illustration of a helical
mandrel for use in this method. The mandrel consists of
a rigid rod 300, shaped into a helix. The mandrel
10   extends longitudinally and circumferentially around a
cylindrical space which defines a core 301 of the
mandrel. In the embodiment shown, the pitch and the
amplitude of the helix are constant along the length of
the mandrel, but they may vary if desired.

15   In order to form a helical portion, a length of
straight flexible tube 1, whose external diameter D_E is
greater than the internal diameter D_M of the core of the
mandrel, is fed generally along the core of the mandrel,
as shown in Figure 12b. Because the tube is wider than
20   the space inside the mandrel, it is forced to adopt a
helical form. The tube may be externally or internally
supported to retain its cross-sectional shape during
this process.

After being treated to make it retain its helical
25   shape, e.g. by thermosetting, the tube is removed from
the mandrel, as shown in Figures 12c and 12d.

As can be seen, the pitch of the helical portion is
the same as the pitch of the mandrel, subject to some
possible relaxation of the tube when removed from the
30   mandrel. The amplitude of the helical portion will be
determined by the external diameter of the tube and the
internal diameter of the core of the mandrel.

The above description concerns a batch processing
method for forming the helical tubing, but this method
35   also lends itself to continuous operation. A continuous
length of flexible tube can be drawn through a
comparatively short length of mandrel, and can be

treated to retain its shape as it is drawn through (for
example, by heating and then cooling a tube formed from
a thermosetting resin).

Experiment has shown that the tube rotates relative
to the mandrel when it is drawn through in this way.
Thus, some form of lubrication may be required to enable
smooth functioning of the process.

Figure 12e is a schematic cross-section through the
tube and the mandrel as the tube is drawn. It will be
seen that the mandrel contacts the outside of the tube,
and so the mandrel can be supported from below (at 320)
without interfering with the drawing process.

The mandrel can be formed in any suitable manner,
and the method of forming the mandrel will depend to a
large extent on the size of the tubes being treated.
The mandrel could be formed by winding a rod around a
member with a circular cross-section, or it may be made
by machining, for example using a CNC milling machine.

Another method of making a graft is described with
reference to Figures 13a and 13b. Figure 13a shows a
straight steel rod 110 held in tension between two
clamps (not shown). A soft steel wire 112 has been
wound on to the steel rod in a helical manner, i.e. to
extend longitudinally and circumferentially of the rod.
The wire 112 is secured in place by silver solder. The
wire 112 forms a guide showing where a tubing 1 is to be
wound around the rod 110, which acts as a mandrel. By
using the wire 112 as a guide, the pitch (or helix
angle) of the tubing when wound onto the rod is
predetermined.

The tubing is then heated and cooled in order to
thermoset it. It is separated from the rod and when it
separates it "relaxes" whereby its helical amplitude
reduces. In this example, the tubing is made of ePTFE.

EXAMPLE 1

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Experiments were carried out using polyvinyl chloride tubing with a circular cross-section. Referring to the parameters shown in Figure 1 the tubing had an external diameter D_E of 12mm, an internal diameter D_I of 8mm and a wall thickness T of 2mm. The tubing was coiled into a helix with a pitch P of 45mm and a helix angle θ of 8°. The amplitude A was established by resting the tubing between two straight edges and measuring the space between the straight edges. The amplitude was determined by subtracting the external diameter D_E from the swept width W:

$$2A = W - D_E$$

So:

$$A = \frac{W - D_E}{2}$$

In this example the swept width W was 14 mm, so:

$$A = \frac{W - D_E}{2} = \frac{14-12}{2} = 1 \ mm$$

As discussed earlier, "relative amplitude" A_R is defined as:

$$A_R = \frac{A}{D_I}$$

In the case of this Example, therefore:

$$A_R = \frac{A}{D_I} = \frac{1}{8} = 0.125$$

Water was passed along the tube. In order to observe the flow characteristics, two needles 80 and 82 passing radially through the tube wall were used to

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inject visible dye into the flow. The injection sites
were near to the central axis 30, i.e. at the "core" of
the flow. One needle 80 injected red ink and the other
needle 82 blue ink.

5   Figure 14 shows the results of three experiments,
at Reynolds numbers R_E of 500, 250 and 100 respectively.
It will be seen in all cases that the ink filaments 84
and 86 intertwine, indicating that in the core there is
swirl flow, i.e. flow which is generally rotating.

10

EXAMPLE 2

           The parameters for this Example were the same as in
Example 1, except that the needles 80 and 82 were
15   arranged to release the ink filaments 84 and 86 near to
the wall of the tubing. Figure 15 shows the results of
two experiments with near-wall ink release, with
Reynolds numbers R_E of 500 and 250 respectively. It will
be seen that in both cases the ink filaments follow the
20   helical tubing geometry, indicating near-wall swirl.
Furthermore, mixing of the ink filaments with the water
is promoted.

           It will be appreciated that this invention, in its
25   first aspect, is concerned with values of relative
amplitude A_R less than or equal to 0.5, i.e. small
relative amplitudes. In a straight tubing portion both
the amplitude A and the relative amplitude A_R equal zero,
as there is no helix. Therefore, with values of
30   relative amplitude A_R approaching zero, the ability of
the tubing portion to induce swirl will reduce. The
lowest workable value of relative amplitude A_R for any
given situation will depend on the speed of flow and the
viscosity and density of the fluid (i.e. Reynolds
35   number) and on the pitch (helix angle) and the
particular use of the tubing portion. Relative
amplitudes of at least 0.05, 0.10, 0.15, 0.20, 0.25,

0.30, 0.35, 0.40 or 0.45 may be preferred.

The various manufacturing methods described herein are not limited to the manufacture of tubing with a relative amplitude equal to or less than 0.5, unless otherwise specified. The methods are considered to be of independent patentable significance and are applicable to the manufacture of tubing with larger amplitudes, whilst also being particularly useful for making tubing of small relative amplitudes.